Effects of Body Weight Support in Running on Achilles Tendon Loading

Authors

Callie Pohlman[1, 3], Andrew Pardee[1, 3], Mikey Friedman[1, 3], Drew Rutherford[1, 3], Charles Nathan Vannatta[4], Thomas W. Kernozek[2]

Affiliations

1 Health Professions, Physical Therapy Program, University of Wisconsin-La Crosse, La Crosse, United States
2 Health Professions, La Crosse Institute for Movement Science (LIMS), La Crosse, United States
3 Sports Physical Therapy, Gundersen Health System, La Crosse, United States
4 La Crosse Institute for Movement Science, University of Wisconsin-La Crosse, La Crosse, United States

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Georg Thieme Verlag, Rüdigerstraße 14,
70469 Stuttgart, Germany

Correspondence

Dr. Thomas W Kernozek
University of Wisconsin-La Crosse
Department of Health Professions
1300 Badger Street
54601 La Crosse
United States
Tel.: 6087858468, Fax: 6087858460
kernozek.thom@uwlax.edu

ABSTRACT

Achilles tendon (AT) tendinopathy is common in runners. Repetitive AT loading may play a role in etiology. Interventions such as body weight support (BWS) may reduce loading on the AT in running. Examine how ground reaction force, AT loading, foot strike, and cadence variables change in running with BWS. Twenty-four healthy female runners free from injury were examined. Participants ran on an instrumented treadmill with and without BWS using a harness-based system at a standardized speed. The system has 4 elastic cords affixed to a harness that is attached to a frame-like structure. Kinematic data and kinetic data were used in a musculoskeletal model (18 segments and 16 degrees of freedom) to determine AT loading variables, foot strike angle, and cadence. Paired t-tests were used to compare each variable between conditions. Ground reaction force was 9.0 % lower with BWS ($p < .05$). Peak AT stress, force, and impulse were 9.4, 11.7 %, and 14.8 % lower when using BWS in running compared to no support ($p < .05$). Foot strike angle was similar ($p < .05$) despite cadence being reduced ($p < .05$). BWS may reduce AT loading and impulse variables during running. This may be important in rehabilitation efforts.

Introduction

Achilles tendon injuries are common in runners with Achilles tendinopathy, accounting for up to 10 % of running injuries [1] Conservative management of tendinopathies often involves exercises that seek to gradually increase load tolerance of tendon tissue. This makes understanding tissue-specific loading an important factor in making management decisions [2]. To this end, various strategies have been employed to reduce or redistribute tissue loads in the lower extremity. Such strategies may include modifying cadence, altering foot strike pattern, or using external support systems for total body unloading. Such strategies may enable clinicians to reduce tissue loads sufficiently to avoid pain, allow training, facilitate tissue recovery, and gradually return to normal participation with activities [3–6]. Body weight support systems and equipment offer promise in reducing loading that can include lower body positive pressure treadmills [7] or either fixed and movable harness systems [8–10]. Harness systems appear to offer a less costly form for clinical use but reductions in loading appear less understood.

The use of BWS during running has been previously shown to reduce vertical loading rates, peak ground reaction forces, knee extensor moments, and patellofemoral joint loading [4, 11–13]. Reducing mechanical loading to injured tendons with exercise therapy appears to be an important consideration for tendon tissue reparative processes [14, 15]. Loads that are excessive in magnitude, duration, or frequency can interfere with adaptation and healing and may be important to AT injury [16]. During the course of treatment, clinicians attempt to match magnitude, volume, and rate of loads to current tendon status and stage of healing based on a response to treatment [6]. Therefore, AT treatment strategies may benefit from our understanding of BWS systems as a load reduction strategy in running.

A harness-based system that uses elastic bands to provide BWS during treadmill-based activities has previously been shown to reduce plantar loading during walking and running as well as ground reaction forces and patellofemoral joint loading in running [13]. To our knowledge, the use of this device to reduce AT loads has not been investigated. Our purpose was to determine how harness-based BWS would influence ground reaction forces, AT-related loading variables (AT stress, force, and impulse), foot strike angle, and cadence in running. It was hypothesized that BWS would reduce GRF and AT-related loading variables, while not affecting cadence and foot strike angle.

Materials and Methods

Subjects

Twenty-four healthy females (age: 25.3 ± 2.2 yrs; mass: 63.2 ± 5.5 kg; height: 169.7 ± 7.2 cm) were recruited. Power calculations to justify the sample size indicated a minimum of 18 participants would be needed to detect changes in Achilles tendon stress based on an alpha of 0.05, correlation of 0.4, and a power of 0.9 from the peak Achilles tendon stress differences in running between rearfoot and non-rearfoot strike patterns based on Lyght et al. [5]. All reported running for fitness 7–10 miles per week, had no history of lower extremity injury or surgery for > 12 months limiting running participation, and had previous treadmill running experience. Each provided informed consent in accordance with the following guidelines [17].

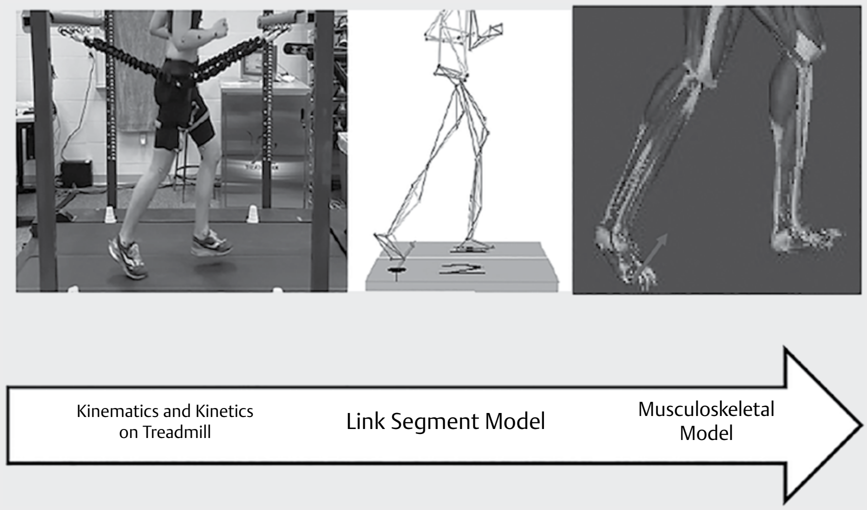
Laboratory procedures

Fifty-four markers were placed on tight-fitting clothing/shoes or the head, trunk, pelvis, and upper and lower extremities [18]. Four markers were placed on the head, 5 on the trunk (C7 and T10 spinous processes, xiphoid process, sternal notch, right scapula), 16 on the bilateral upper extremities (acromion, deltoid muscle, medial and lateral humeral epicondyles, forearm, ulnar and radial styloid processes, and the second metacarpophalangeal joint), 5 on the pelvis (both anterior superior iliac spines and both posterior superior iliac spines, and the apex of the sacrum), and 16 on both lower extremities (greater trochanter, anterior thigh, lateral femoral epicondyle, anterior tibia, lateral malleolus, heel of the shoe, 2nd and 5th metatarsophalangeal joint) [18]. All wore the same shoe type (Zealot; Saucony, Boston, MA, USA) due to shoe-related differences that may influence running performance characteristics. Kinematics were collected at 180 Hz using a 12-camera motion capture system (Motion Analysis Corp., Santa Rosa, CA, USA). Kinetics were collected at 1800 Hz using a split belt instrumented treadmill (Treadmetrix, Park City, UT, USA). Kinematic and analog data from the treadmill force plate were filtered with a 15 Hz low-pass Butterworth filter [19].

The BWS system (Lightspeed Lift, Duluth, MN, USA) uses an external frame surrounding the treadmill where elastic cords are affixed to a harness worn by each participant (▶**Fig. 1**). The elastic nature of the cords coupled with their respective angle of pull provide an upward force on the runner. A harness was placed on the legs and waist per the manufacturer's recommendation and a 3-minute running treadmill warm-up was performed at 3.3 m/s (8.1 min/mile pace, average of this study cohort). In a randomized order, participants ran under two conditions: control (no BWS) and using an attachment point that attached the elastic cords 30 cm above each participant's greater trochanter when standing from the frame to the harness. The 30-cm attachment point was based on manufacturer guidelines in their operations manual. Each participant ran 4 minutes for each condition, where data were obtained from the last 30 seconds. Approximately a 2-minute rest period allowed for BWS adjustments between conditions. Before the running trials, images of the AT were acquired using a GE LOGIQ Ultrasound P6 (Waukesha, WI, USA) with a ML6–15 probe. Participants were positioned prone on a treatment table with their right ankle measured to 90 ° in a neutral position with a goniometer. The position was chosen to avoid the anisotropy effect by facilitating contact between the probe and the tendon [20]. Ultrasound gel (Aquasonic Clear, Fairfield, NJ, USA) was applied to the head of the probe. Transverse images of the AT were collected by placing the probe 10 cm proximal to the calcaneal insertion on the posterior aspect of the shank between the medial and lateral malleoli perpendicular to the AT. Ultrasound AT cross-sectional areas were measured using ImageJ (Wayne Rasbrand, National Institutes of Health, USA) software.

Data analysis

Using a musculoskeletal model (Human Body Model; Motekforcelink, Amsterdam, Netherlands) with 16 segments and 46 total degrees of freedom (DOF), muscle forces were determined. The trunk had 3 segments: pelvis, midtrunk, and thorax, each with 3 DOF. Shoulders had 6 DOF relative to the thorax and elbow and wrist joints each had 2 DOF. The pelvis had 6 DOF, hip with 3 DOF, knee with 1 DOF and subtalar and ankle joint each were modeled with 1 DOF. The knee and ankle joint centers were 50 % of the joint width and marker diameter from a static pose. Bell et al. [21] was used to determine hip center from pelvis markers.. The Levenberg-Marquardt algorithm was used to minimize the musculoskeletal model pose [18] and model kinematics were solved using a global optimization approach. Inertial properties of body segments were based on De Leva et al. [22]. Three hundred muscle tendon units were represented in this model where the muscle parameters, insertion points, and wrapping points were based on Delp et al. [23]. Respective to the DOF allowed at each joint, muscle forces were estimated using static optimization such that the sum of squared muscle forces were minimized relative to the muscles' maximum strengths matching the measured joint moments from inverse dynamics [18]. To solve the quadratic programming issue, a recurrent neural model was used [24]. Viscoelastic properties of the muscles were not modeled in this musculoskel-



▶ **Fig. 1** Top left photo of the BWS system with a participant while running on the instrumented treadmill with motion analysis markers. During the control condition, the participant ran with the harness but the elastic cords were not attached to the frame of the lift. This figure provides a visual representation of the steps used to determine the Achilles tendon (AT) loading during running.

etal model. Muscle forces have been reported similar to Opensim during gait performance when used with similar modeling parameters [25]. Data were used in a custom AT model to determine AT force [26]. Foot strike angles were calculated as the angle between the vector connecting the marker on the heel and the 2nd digit of foot along the anteroposterior axis in the lab coordinate system using a custom MATLAB script for each step (MathWorks, Inc., Natick MA, USA) based on Altman & Davis [27].

A 50 N vertical force threshold was used to determine the right leg stance for 8 successive steps. The mean of these steps was determined for AT-related loading variables (AT force and impulse), vertical GRF, foot strike angle, and cadence.

Statistical analysis

A multivariate analysis with repeated measures examined for differences in peak vGRF, peak AT stress, peak AT force, AT impulse, foot strike angle, and cadence between control and the BWS condition in SPSS 28.0 (IBM Corporation, Armonk, NY, USA). Alpha was set to 0.05. Effect sizes were calculated using Cohen's *d*.

Results

▶ **Table 1** depicts the descriptive statistics, p-values, and effect sizes of each dependent variable with and without BWS. Multivariate analysis revealed collective differences between the control and BWS condition (Wilk's Lambda $p<0.001$). Univariate tests indicated there were differences in peak vGRF. ▶ **Fig. 2** depicts the ensemble averaged vGRF and AT stress for the control and BWS condition during the stance phase of running. Peak vGRF was reduced 9.0 % for the BWS condition. Peak AT stress was reduced by 9.4 % with the use of the BWS. Peak AT force and AT impulse were reduced for the BWS condition. BWS reduced loading in these variables 11.7 and 14.8 %, respectively. Foot strike angle was not different ($p > 0.05$) with BWS despite increasing 2.2 % while cadence decreased by 3.4 % with BWS ($p < 0.05$).

Discussion

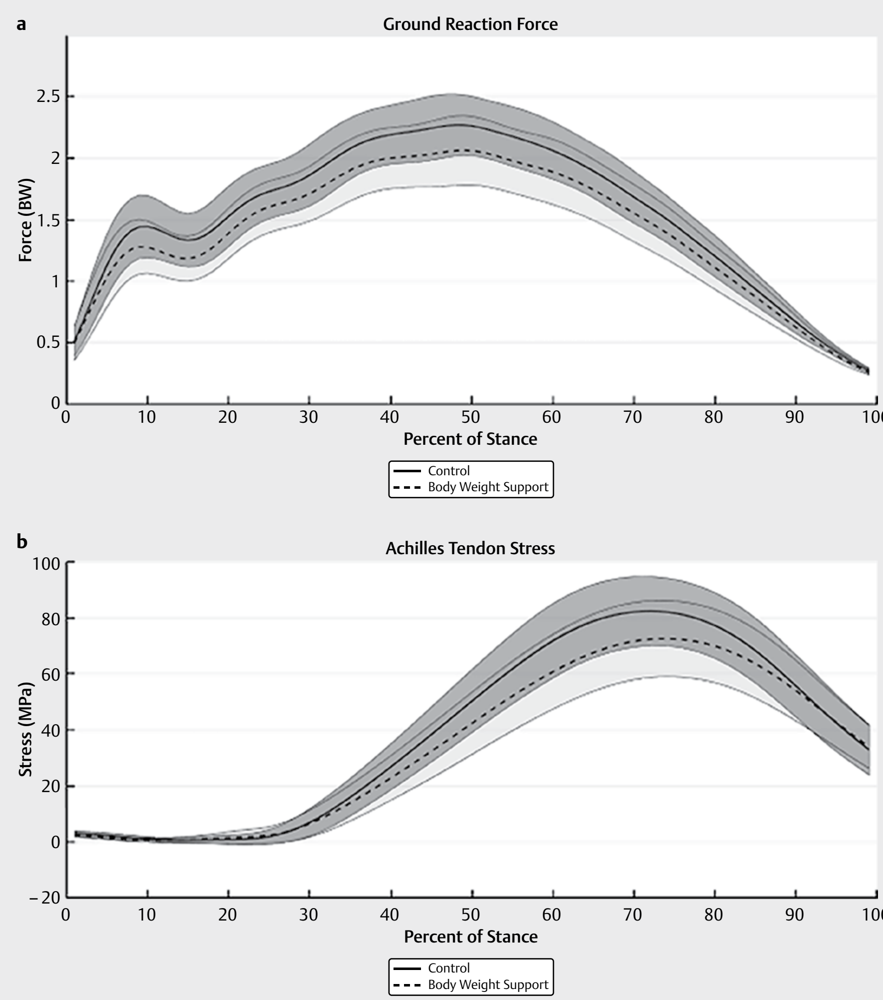
Our investigation examined how BWS would influence ground reaction forces, AT-related loading variables (AT stress, force, and impulse), foot strike angle, and cadence in running. All variables were reduced with the use of BWS except for foot strike angle, which exhibited no change with BWS.

The measures of AT stress and force were higher in this study than those in previous studies. This cohort demonstrated a peak AT stress of 83.25 MPa in the control condition compared to previous works reporting a peak AT stress of 56.9 and 69.9 MPa [5, 26]. Similarly, peak AT force was higher in the present study than in previous works. This cohort of runners demonstrated peak AT force of 7.51 BW while previous work reported peak AT force in rearfoot strike running at similar speeds of 6.46 and 5.6 BW [5, 28]. A potential explanation for the higher values in the present study is that data collection occurred on a treadmill compared to the other studies all being completed during overground running. When comparing treadmill running to overground running, a 16.6 % increase in plantarflexion moment along with an increase of 29.75 % power absorption at the ankle has been reported during treadmill running [29].

Peak AT stress, force, and impulse were all reduced with the BWS. Previous work has shown that increasing cadence may also reduce AT loading regardless of foot strike pattern. For example, Lyght et al., [5] reported that increasing cadence by 5 % from a preferred cadence reduced AT stress by 4.2 and 2.9 % in rearfoot and forefoot strike conditions, respectively. Considering that the use of BWS achieved a 9.4 % reduction in AT stress, it seems that BWS may lead to greater reductions in AT stress than altering cadence.

▶ **Table 1** Mean and standard deviation for vertical ground reaction force, Achilles tendon (AT) loading variables, foot strike angle, and cadence with and without BWS (control). P-values and effect sizes depict differences between the BWS and control condition.

	Control	**Body Weight Support**	
Variable	**Mean (sd)**	**Mean (sd)**	**p-value, Cohen's *d***
Peak Ground Reaction Force (BW)	2.34 (0.26)	2.12 (0.29)	$p<0.0001$, $d=2.38$
Peak Achilles Tendon Stress (MPa)	83.25 (12.54)	73.43 (14.34)	$P<0.0001$, $d=0.73$
Peak Achilles Tendon Force (BW)	7.51 (1.13)	6.63 (1.30)	$p<0.0001$, $d=1.52$
Achilles Tendon Impulse (BW * s)	0.82 (0.16)	0.70 (0.17)	$p<0.0001$, $d=1.94$
Foot Strike Angle (°)	26.65 (4.35)	27.25 (4.51)	$p=0.133$, $d=-0.23$
Cadence (Steps/minute)	169.13 (8.36)	163.63 (9.22)	$p<.0001$, $d=1.91$



▶ **Fig. 2** **a**) Ensemble average and standard deviation of the peak vertical ground reaction force (body weight) for both conditions (control and body weight support (BWS)) as a percentage of the stance phase of running. **b**) Ensemble average and standard deviation of the peak Achilles tendon stress (MPa) for both conditions (control and BWS) as a percentage of stance.

Similar to changes seen with AT stress, BWS also resulted in relatively large reductions in peak AT force. Previous work has demonstrated that a 5 % increase in cadence resulted in a 3.6 and 2.7 % reduction in AT force for rearfoot strikers and forefoot strikers, respectively. The present study demonstrated an 11.7 % decrease in peak AT force in rearfoot strikers. Thus, in patients with symptomatic Achilles tendinopathy who do not achieve symptom reduction with cadence manipulation, BWS may provide a reasonable therapeutic alternative to reducing AT load in running.

There was also a large reduction in peak AT impulse using the BWS. This reduction in impulse influences the cumulative load on the AT during stance and therefore may have larger implications on the magnitude of cumulative loading on the AT over the course of a training run. For example, in the control condition over the course of running one kilometer, runners would impose a cumulative load of 693.6 BW · s/km to the Achilles tendon while running under BWS conditions would only impose a cumulative load of 571.1 BW · s/km. This amounts to a 17.7 % reduction in the cumulative load per kilometer. Thus, the use of BWS may provide a means to reduce the cumulative AT load in those runners that utilize a rearfoot strike pattern.

Although our study did not directly manipulate cadence, we found that using BWS decreased cadence. The pattern of increased BWS with decreased cadence has been observed in other studies. Masumoto et al. [30] reported that increasing BWS by 50 and 80 % decreased preferred stride frequency by 10.5 and 14.6 %, respectively. However, AT loading variables were not examined within their investigation. Because previous work has demonstrated that decreasing cadence is accompanied by increased AT stress and force [5], it is reasonable to conclude that the reduction in AT loading variables observed in this study are the result of BWS and not due to changes in cadence.

The peak vGRF with BWS was reduced by 9.0 % compared to the control. Grabowski and Kram [30] also concluded that using a lower body positive pressure treadmill reduced peak impact GRF linearly while running at three different velocities (3.0, 4.0, and 5.0 m/s). It is likely that many of the changes in AT loading are driven by the overall reduction of impacts during running.

It should be noted that running speed has been shown to reduce AT loading variables and that simply reducing speed from 3.9 m/s to 3.3 m/s that peak AT force and cumulative AT loading/kilometer reduced 2.67 and 3.4 %, respectively [28]. However, often the intent with interventions such as cadence manipulation or BWS systems is to allow runners to continue to train at an intensity that would allow for continued cardiovascular training to occur. Although reducing running speed would lead to less metabolic demand, the use of BWS systems to unload runners has also been shown to reduce metabolic demand at a given running speed [31]. Implementation of BWS may need to consider the tradeoff between metabolic demands and tissue loading when implementing a rehabilitation strategy for AT-related injury.

AT-related injuries are common in runners due to the large forces and repetitive loads applied to the lower extremities [32]. Running with a rearfoot strike pattern has the potential for decreasing the risk of injury due to the smaller loads placed on the AT [3]. As no concurrent changes were seen in foot strike angle with BWS, this may indicate that runners may be able to maintain their typical foot strike pattern while still benefiting from reduced loading to the AT when using this system. This may be useful for rehabilitation and offloading while using a typical running pattern.

BWS may also be beneficial for orthopedic conditions such as AT injuries or rehabilitation after rupture. Saxena and Granot [33] examined the effectiveness of using a positive pressure treadmill to obtain body weight support in the return to activity following AT injury. The BWS group were able to return to running nearly two weeks faster than the control group. The progressions of dosing BWS for running included 70 % BW at 13.9 weeks, 85 % at 17.6 weeks, and 100 % BW at 18.1 weeks. This study demonstrated that BWS can be used for walking and running but also for neuromuscular reeducation and during concentric strengthening exercises [33]. The cost of anti-gravity treadmills is quite expensive compared to harness-based systems, however the precise magnitude of offloading may be less precise [13]. At present, the amount of load reduction dosage for various running-related injuries is largely unknown despite offloading systems potential in being useful to rehabilitation. Nonetheless, harness-based systems appear to offer a means of reducing tissue-specific loads, which may enhance rehabilitation efforts.

Limitations

There are several limitations of this study. First, there is not a precise quantification of amount of BW offloaded based on a set height of the harness used. The elastic nature of the bungee cord system with individuals of varying body weight and vertical oscillation influences the magnitude of offloading. The correlation between body weight and maximum ground reaction force in this investigation was r = 0.878, indicating that 76.7 % of the variance in body weight explained the variation in peak vertical ground reaction force. Additionally, plastic deformity of the bungee cords over time may have also contributed to subtle changes in loading for participants tested earlier compared to later. Thus, precise quantification of percentages of body weight may not be possible with this system, but it appears effective in unloading tissues of the lower extremity. Next, musculoskeletal modeling approaches have a variety of anatomical and mechanical assumptions. However, many of these modeling assumptions may not have influenced study findings as a repeated measures design was used for this experiment. This study investigated only rearfoot strike runners. How the AT is unloaded in non-rearfoot strike runners is unknown. Lastly, changes in loading rates were not quantified in this study, but it should be acknowledged that loading rates are an important consideration in load management programs.

Conclusion

Overall, a harness-based BWS system was effective in reducing vGRF, AT loading variables, and cadence while not inducing changes in foot strike pattern compared to no body support during running. Use of this harness-based system may be useful in rehabilitation when reduction in AT loading is desired.

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Conflict of Interest

The authors declare that they have no conflict of interest.

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